The effect of overcommitment and reward on muscle activity, posture, and forces in the arm-hand-wrist region - a field study among computer workers

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Stepping strategies used by post-stroke individuals to maintain margins of stability during walking

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ABSTRACT

Background: People recovering from a stroke are less stable during walking compared to able-bodied controls. The purpose of this study was to examine whether and how post-stroke individuals adapt their steady-state gait pattern to maintain or increase their margins of stability during walking, and to examine how these strategies differ from strategies employed by able-bodied people.

Methods: Ten post-stroke individuals and 9 age-matched able-bodied individuals walked on the Computer Assisted Rehabilitation Environment. Medio-lateral translations of the walking surface were imposed to manipulate gait stability. To provoke gait adaptations, a gait adaptability task was used, in which subjects occasionally had to hit a virtual target with their knees. We measured medio-lateral and backward margins of stability, and the associated gait parameters walking speed, step length, step frequency, and step width.

Findings: Post-stroke participants showed similar medio-lateral margins of stability as able-bodied people in all conditions. This was accomplished by a larger step width and a relatively high step frequency. Post-stroke participants walked overall slower and decreased walking speed and step length even further in response to both manipulations compared to able-bodied participants, resulting in a tendency towards an overall smaller backward margins of stability, and a significantly smaller backward margin of stability during the gait adaptability task.

Interpretation: Post-stroke individuals have more difficulties regulating their walking speed, and the underlying parameters step frequency and step length, compared to able-bodied controls. These quantities are important in regulating the size of the backward margin of stability when walking in complex environments.

1. Introduction

People who are recovering from a stroke have an increased risk of falling during walking (Weerdesteyn et al., 2008). In the literature several causes for this increased risk of falling are suggested, such as an enlarged body sway in the fontal plane during steady state walking (De Bujunda et al., 2004; Tyson, 1999), and a limited capacity to adapt the gait pattern in response to environmental demands, for example to avoid an obstacle (Den Otter et al., 2005; Said et al., 1999, 2008). Especially when obstacles suddenly appear and fast and accurate adaptations are necessary, the failure rate in post-stroke individuals is higher compared to able-bodied people (Den Otter et al., 2005). Besides, not only the higher probability of an obstacle collision, but also an impaired postural stability during and after obstacle crossing might increase fall risk in post-stroke individuals (Said et al., 2008).

Fall risk during walking can be assessed by determining the margin of stability (MoS). The MoS is defined as the distance between the extrapolated centre of mass (XCoM) and the limits of the base of support, in which the XCoM is a concept that takes both the position and the velocity of the centre of mass (CoM) into account (Hof et al., 2005). The MoS can be calculated in both medio-lateral (ML) (Hof et al., 2007; McAndrew Young & Dingwell, 2012; McAndrew Young et al., 2012) and anterio-posterior (AP) (Espy et al., 2010b; McAndrew Young & Dingwell, 2012; McAndrew Young et al., 2012; Pai & Patton, 1997 Apr) direction, in which the AP MoS is usually calculated with respect to the base of support (BoS) of the leading foot at initial contact. The difficulty with the interpretation of the AP MoS is that an increase in AP MoS in backward direction by definition implies a decrease of the AP MoS in forward direction. However, from previous experiments we know that, when balance is threatened, people prioritize an increase in backward (BW) MoS, limiting the chance of a backward loss of balance, above an increase in forward MoS (Bierbaum et al., 2010, 2011; Hak et al., 2012; McAndrew Young et al., 2012). MoS can be regulated effectively by adjusting step parameters. From studies of Hof

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et al. (Hof, 2008; Hof et al., 2005, 2007), it appeared that possible strategies to increase the ML MoS during walking are an increase in step width and step frequency, while Espy et al. (Espy et al., 2010a, 2010b) have shown that a decrease in step length and an increase in walking speed have a positive effect on the size of the BW MoS.

To assess the risk of falling, ML and BW MoS can be measured during unperturbed walking. However, measuring the MoS during more challenging walking conditions allows one to investigate whether subjects are able to use active adjustments of the gait pattern to increase or at least maintain the ML and BW MoS. Previous studies have found that not only able-bodied people, but also people who walk with a trans-tibial prosthesis successfully exploit such strategies. In response to continuous platform perturbations they increase step width and step frequency, resulting in an increase in ML MoS, and they decrease step length while keeping walking speed constant, resulting in an increase in BW MoS (Hak et al., 2013b; Hak et al., 2012; McAndrew Young et al., 2012). In other recent studies we investigated whether able-bodied people and trans-tibial amputees were able to control their MoS during a task in which besides maintaining gait stability, fast and accurate adaptations of the gait pattern had to be made, to hit virtual targets with the knees (Hak et al., 2013a,b). The available response time was very short, because targets appeared within the same stride as they had to be hit. We found that both subject groups decreased step length and increased in step width in their average gait pattern. These adaptations appeared to be mainly an anticipatory strategy to facilitate the fast and accurate response necessary to hit the targets, and to prevent a loss of balance while performing the task (Hak et al., 2013a). Simultaneously, no increase in step frequency was found in this situation, probably to prevent a further decrease of the available response time which would hamper an accurate adaptation (Hak et al., 2013a,b).

During unperturbed walking, the gait pattern of post-stroke individuals already differs from the gait pattern utilized by able-bodied people and some aspects of this deviant gait pattern have been explained as mechanisms to regulate gait stability (Chen et al., 2005; Krasovsky et al., 2012). In the study of Chen et al. (Chen et al., 2005), the larger step width in post-stroke individuals was explained as a compensation for the larger body sway in the frontal plane. A lower walking speed in people with gait impairments is frequently explained as a strategy to increase gait stability (Dingwell & Marin, 2006; England & Granata, 2007; Kang & Dingwell, 2008; Krasovsky et al., 2012). However, a lower walking speed may decrease the BW MoS (Espy et al., 2010a; Pai & Patton, 1997). Besides, when a reduced walking speed coincides with a decrease in step frequency it may also have a negative effect on the size of the ML MoS (Hof et al., 2005; Hof et al., 2007). Hence, it is unknown whether and how changes in the steady state gait pattern of people who have suffered from stroke affect their MoS and whether people after stroke can adapt their steady state gait pattern to increase or preserve their MoS during challenging walking conditions. Therefore in the current study we manipulated gait stability and gait adaptability during walking (Hak et al., 2012; Hak et al., 2013a,b). We assessed whether post-stroke individuals use similar strategies as able-bodied people to preserve MoS during unperturbed walking and when required to withstand manipulations of gait stability or to facilitate gait adaptability. We hypothesized that post-stroke individuals walk with smaller MoS, compared to the able-bodied controls, and that MoS decreased even further, for the post-stroke individuals, during the manipulations of gait stability and adaptability. The main reason for these differences in MoS between both groups might be the lower walking speed, which will influence the size of the BW MoS negatively, and the lower step frequency which will decrease the ML MoS.

2. Methods

2.1. Subjects

Ten adult subjects who had suffered from a stroke (age mean 60.8 (SD 8.4) years, height mean 1.79 (SD 0.07) m, mass mean 88.4 (SD 8.5) kg) and 9 age-matched control subjects (age mean 57.3 (SD 7.2) years, height mean 1.77 (SD 0.08) m, mass mean 79.7 (SD 9.0) kg) participated in this study. Post-stroke participants and able-bodied controls were respectively recruited from the patient population and the employees of the Military Rehabilitation Centre Aardenburg, Doorn, The Netherlands. A minimum score of 4 on the Functional Ambulation Categories (FAC) (Holden et al., 1986) in combination with a minimum score of 45 on the Berg Balance Scale (BBS) (Stevenson, 2001) was required to participate in this study. Further characteristics of the post-stroke group are reported in Table 1. This study was approved by the medical ethical committee (Ref: NL35402.029.11) and all subjects gave their written informed consent in accordance with university policy.

2.2. Equipment

All subjects walked in the Computer Assisted Rehabilitation (CAREN, Motek Medical b.v., Amsterdam, The Netherlands), which consists of an instrumented treadmill mounted onto a 6-degree-of-freedom motion platform in combination with a Virtual Environment (VE) (Fig. 1A). Twelve high resolution infra-red cameras (Vicon, Oxford, UK) were used to capture kinematic data of 16 reflective markers attached to pelvis and the lower extremities (lower body plug-in-gait (Davis et al., 1991; Kadaba et al., 1990)). The treadmill was used in the self-paced mode, which allowed subjects to modify walking speed at will. This was done by servo-controlling the motor with a real-time algorithm that took into account the pelvis position in the AP-direction, as measured by the markers attached to the pelvis, and a reference position on the treadmill, corresponding to the AP-midline of the treadmill. A safety harness system suspended overhead prevented the subjects from falling, but did not provide weight support.

2.3. Protocol

2.3.1. Familiarization

Before the protocol started, subjects performed at least 5 familiarization trials of 3 min each, to become familiar with walking on a (self-paced) treadmill, the VE and the various manipulations.

2.4. Experimental trials

The actual protocol consisted of 3 trials of 4 minute walking at self-paced walking speed: 1) a trial of unperturbed walking, 2) a trial with a continuous perturbation of the motion platform, and 3) a trial with a gait adaptability task. The first minute of each trial was used to let subjects get used to the self-paced setting of the treadmill and the manipulation concerned. All trials were offered in random order.

For the platform perturbation, translations of the walking surface in ML-direction were used, following a multi-sine function (Hak et al., 2012; McAndrew et al., 2010a, 2010b) (Fig. 1B).

For the gait adaptability task (GA task) the VE was used to project targets on the screen (Fig. 1C). In addition, a projection of the markers attached to the knees was shown on the screen. Subjects were instructed to hit the targets on the screen with the projected knee markers that were attached to the lateral epicondyles, as close as possible to the centre of the targets. The purpose of this task was to simulate a situation that requires accurate and fast adaptations of the normal stable gait pattern, with a limited response time, for example to avoid an obstacle that suddenly appears. A reason to choose for this specific task instead of virtual obstacle avoidance tasks is the

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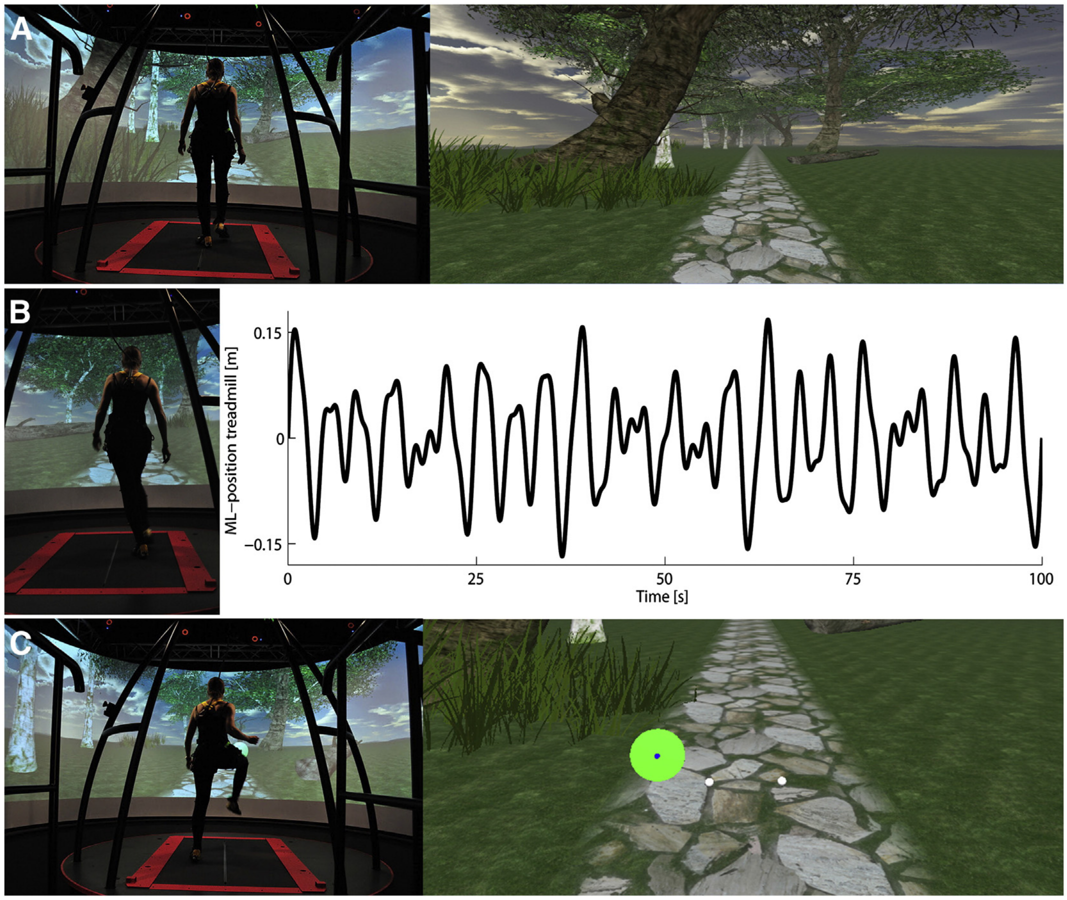

Fig. 1. (A) Experimental setup: CAREN and Virtual scene; (B) GA-task with an example of a target and the projection of the knee markers in the right panel; (C) ML balance perturbation with the perturbation pattern in the right panel.

impossibility to step over virtual objects that are projected on a 2D screen. Stepping over virtual objects would require a 3D environment. Another advantage of the adaptability task used in the current study is the possibility to quantify the performance on the task in terms of accuracy of the knee movement, while for an obstacle avoidance task only a pass and a hit can be distinguished from each other to quantify the performance. Within this trial, a total of 32 targets appeared with a time interval of about 5 s in between. Targets appeared at initial contact and disappeared after approximately one gait-cycle, the duration of which was estimated in the first minute of the trial. The positions of the targets differed randomly in side (left or right), height (120% or 140% of lower leg length), and ML-position (120% or 140% of distance between the left and right anterior superior iliac spines from the midline of the treadmill), to increase the unpredictability of this task (Hak et al., 2013a).

2.5. Data collection

Kinematic data of markers attached to the lateral malleoli of the ankles, the pelvis (left and right anterior superior iliac spines (LASI & RASI), and left and right posterior superior iliac spines (LPSI & RPSI)), and the lateral epicondyle of the knees were collected with the Vicon system. The sample rate of data collection was 120 samples/s. The final three minutes of each trial were used for data analysis. Before data analysis, both speed data and kinematic data were low-pass filtered with a 4th order bi-directional Butterworth filter with a cut-off frequency of 10 Hz.

2.6. Data analysis

All outcome measures, walking speed, the different step parameters and the margins of stability, were averaged across the total numbers of steps during the final 3 min of each trial. However, before calculating these outcome measures for the trials with GA-task, we removed the strides in which the targets had to be hit, to focus on the anticipatory strategy to facilitate the gait adaptations.

2.7. Walking speed

Walking speed was calculated as the average treadmill speed over the final 3 min of each trial.

2.8. Step parameters

Step frequency was determined as the inverse of the average duration between two subsequent heel-strikes, where heel-strikes were detected as the local maxima of the position of the ankle markers in the AP-direction. Step width was calculated as ML-distance between both ankle markers at the instant of heel-contact and step length was defined as the AP-distance between these markers at the instant of heel-contact.

2.9. Gait stability

To quantify gait stability, margins of stability (MoS) in medio-lateral (ML) and backward (BW) direction were calculated, following a method derived from the method introduced by Hof (Hof et al., 2005, 2007), as the difference in ML and BW direction between the extrapolated centre of mass (XCoM) and the margin of the base of support (BoS) (Fig. 2). The XCoM is a concept that takes both the position and velocity of the centre of mass (CoM) into account. MoS were calculated for the instant at which the MoS reached its minimum value within each step, which is always at the instant of initial contact for the BW MoS.

Our method is basically similar to that of Hof (Hof et al., 2005 2007) who used force plate data for calculating the trajectory of the CoM and the XCoM. The difference with the current study is that the average of the pelvis markers was used to estimate the position of the CoM. The markers attached to the ankles were used to define the margin of the BoS.

To quantify the amount of body sway in the frontal plane the maximal difference in XCoM between subsequent steps was calculated (ML $XCoM_{disp}$; Fig. 2A). This method adds to the method used in previous studies (De Bujunda et al., 2004; Tyson, 1999), in which only the displacement of the CoM was used to calculate the amount of body sway during walking. By taking the XCoM also the velocity of the CoM was taken into account.

To differentiate between the contribution of step length and walking speed on the size to the BW MoS, we additionally calculated the distance between the BoS and the CoM in backward direction (BW BoS-CoM_{dist}; Fig. 2B) and the forward velocity of the CoM (FW vCoM) at initial contact. Resulting values were averaged over steps.

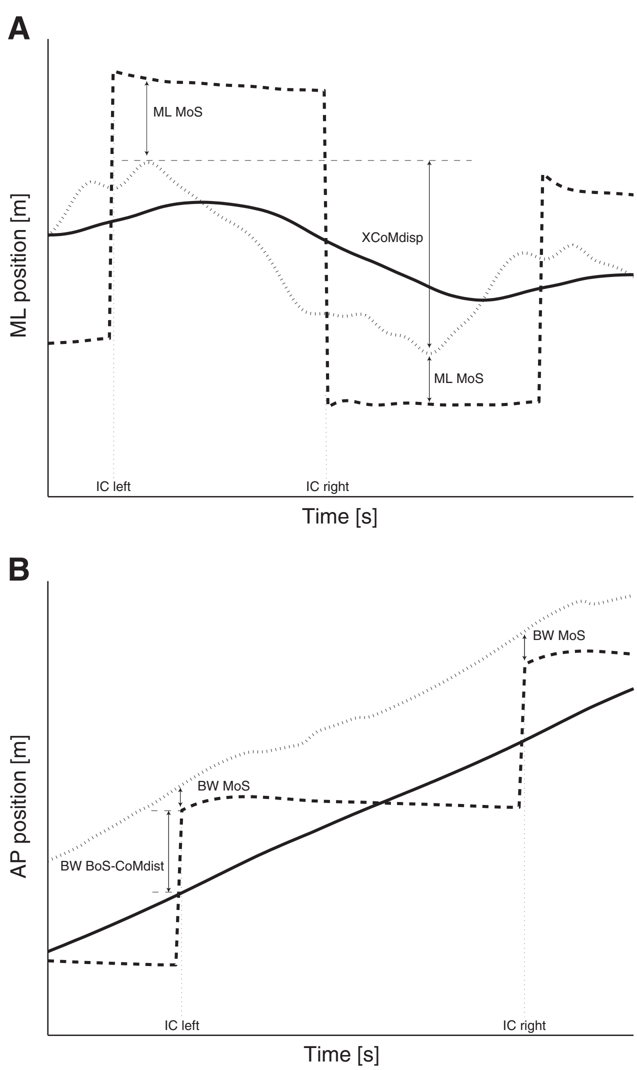

Fig. 2. Schematic representation of the calculation of (A) the ML MoS, the ML $XCoM_{disp}$, and (B) the BW MoS and the BW BoS-CoM_{dist}. Trajectories of the margin of the BoS (dashed line), CoM (solid line), and XCoM (dotted line) are shown for a period of approximately 2 steps. The XCoM is calculated as the position of the CoM plus its velocity times a factor $\sqrt{(l/g)}$, with l being the maximal height of the origin of the pelvis and g the acceleration of gravity. The MoS is the difference between the trajectory of the XCoM and the margin of the BoS at the instant at which the MoS reached its minimum value within the period of one step for the ML MoS and at initial contact for the BW MoS. In ML direction the $XCoM_{disp}$ was calculated for the same instant at which the ML MoS was calculated. The BW BoS-CoM_{dist} was calculated at initial contact.

2.10. Gait adaptability

Gait adaptability was quantified by the performance on the GA-task. This performance is defined as the minimum distance between knee and target centre. For the period in which the target was visible on the screen the minimal Euclidean distances between the knee markers and the center of the target in the plane of projection of the VE were assessed for each projected target. The average of these distances was calculated.

2.11. Statistical design

To determine the effects of the platform perturbation and the GA-task on step length, step frequency, step width, walking speed, ML $XCoM_{disp}$, and ML and BW MoS, and to investigate whether these effects differ between post-stroke participants and able-bodied people, 2 × 3 factorial ANOVAs were performed. The three conditions (normal walking, perturbed walking and walking with GA-task) were used as within factor and group as between factor. Simple contrasts were used to determine whether outcome measures differed between normal walking and either perturbation task or adaptability task, and whether there were group by condition interaction effects. *P*-values less than 0.05 were considered significant. In the case of a significant interaction effect, paired-samples t-tests with a Bonferroni correction (critical *P*-value: 0.025) were performed to investigate for each group separately whether the parameter concerned was affected by the manipulation. A Mann–Whitney *U* test was used to investigate whether the performance on the GA-task differed between groups. For this comparison, a non-parametric test was chosen because the distribution of the performance scores was not normal for the post-stroke group. Also for the outcome of the Mann–Whitney *U* test, a *P*-value less than 0.05 was considered significant. Statistical analyses were performed using IBM SPSS Statistics 20.0.

3. Results

One of the participating post-stroke participants (subject number 5, Table 1) was not able to complete the trials with the perturbation and the GA-task. Therefore, the analyses were achieved in nine post-stroke participants and nine able-bodied controls. A visual representation of the results for the gait parameters (walking speed, step frequency,

Table 1
Characteristics stroke patients.

Subject#	Cognitive disturbances	Side hemiparesis	Berg Balance Scale	Time since stroke (months)
1	None	Left	56	8
2	Minor	Left	51	10
3	Aphasia	Left	52	8
4	None	Left	51	4
5[a]	None	Right	47	9
6	Moderate	Left	47	38
7	None	Right	52	1
8	Aphasia	Right	56	1
9	None	Right	45	1
10	Minor	Left	55	12

[a] Excluded from data analyses.

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step length and step width) and the margin of stability measures is presented in Figs 3, 4 and 5. Results of the statistical analyses are shown in Table 2.

A significantly smaller step length, a decreased walking speed, and a larger step width were observed in the post-stroke group compared to the controls. Step frequency did not differ significantly between both groups. Also for the ML and BW MoS no significant group effects were found, although the difference in BW MoS between groups nearly reached the level of significance ($P = 0.077$), with smaller margins for the post-stroke participants. The $XCoM_{disp}$ was significantly larger and BW $BoS\text{-}CoM_{dist}$ and FW vCoM were significantly smaller for the post-stroke group compared to the able-bodied controls.

A main effect of the platform perturbation on all recorded gait parameters was found. Subjects increased step width and step frequency and decreased step length and walking speed in response to the perturbation. For step length and walking speed, group ∗ perturbation interactions were found. Post hoc analyses showed that step length decreased in both groups, but this decrease was larger in the post-stroke group. Walking speed decreased only in the post-stroke group. In response to the GA-task, step length and walking speed decreased, step width increased, while step frequency was not significantly affected by the GA-task. For walking speed, a group∗GA-task interaction effect was found. Post hoc analyses showed that in response to the GA-task walking speed decreased only in the post-stroke group.

In response to the platform perturbation, both ML $XCoM_{disp}$ and the ML MoS increased in both groups. BW MoS was not affected significantly, while BW $BoS\text{-}CoM_{dist}$ and FW vCoM decreased in both groups. For FW vCoM and BW $BoS\text{-}CoM_{dist}$, also significant group ∗ perturbation interactions were found. Post-hoc test showed that FW vCoM only decreased for the post-stroke group ($P < 0.01$ for the post-stroke group; $P = 0.204$ for the able-bodied controls). BW $BoS\text{-}CoM_{dist}$ decreased in both groups ($P < 0.01$ for both groups), but this decrease was larger in the post-stroke group.

In response to the GA-task, neither the BW MoS nor the ML MoS was affected significantly. However, a significant group∗GA-task interaction effect for the BW MoS was found. Post-hoc analyses showed that BW MoS significantly increased in response to the GA-task in the able-bodied participants ($P < 0.01$), while BW MoS slightly decreased in the post-stroke group, however this was not significant ($P = 0.104$). ML $XCoM_{disp}$ increased significantly, while BW $BoS\text{-}CoM_{dist}$ and FW vCoM decreased significantly in response to the GA-task. For both BW $BoS\text{-}CoM_{dist}$ and FW vCoM also a significant group ∗ GA-task interaction effect was found. Post-hoc analyses showed that both BW $BoS\text{-}CoM_{dist}$ and FW vCoM only decreased in the post-stroke group.

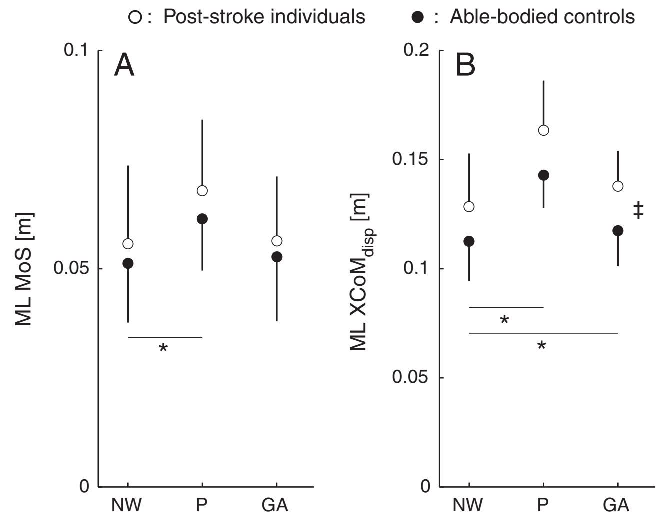

Fig. 4. Average and standard deviation of ML MoS (A) and ML $XCoM_{disp}$ (B) for post-stroke individuals (n = 9) and healthy controls (n = 9). NW: normal walking; P: perturbation; GA: gait adaptability task. Significant group effects are indicated with ‡. Significant contrasts between normal walking and perturbed walking and/or between normal walking and walking with GA-task are indicated with *. For these data no significant groupx manipulation interaction were found.

The performance on the GA-task, quantified as the average distance between knee and target, was mean 13.02 (SD 18.04) cm in the post-stroke participants and mean 3.74 (SD 1.02) cm in the able-bodied participants, and differed significantly between both groups ($U = 77.00$; $P < 0.001$; $df = 16$).

4. Discussion

The purpose of the current study was to evaluate whether post-stroke individuals preserve their margins of stability during walking with manipulations of gait stability and gait adaptability and whether post-stroke individuals use similar strategies to withstand perturbations of gait stability or to facilitate gait adaptability as able-bodied people. For the trials of unperturbed walking and for walking with the platform

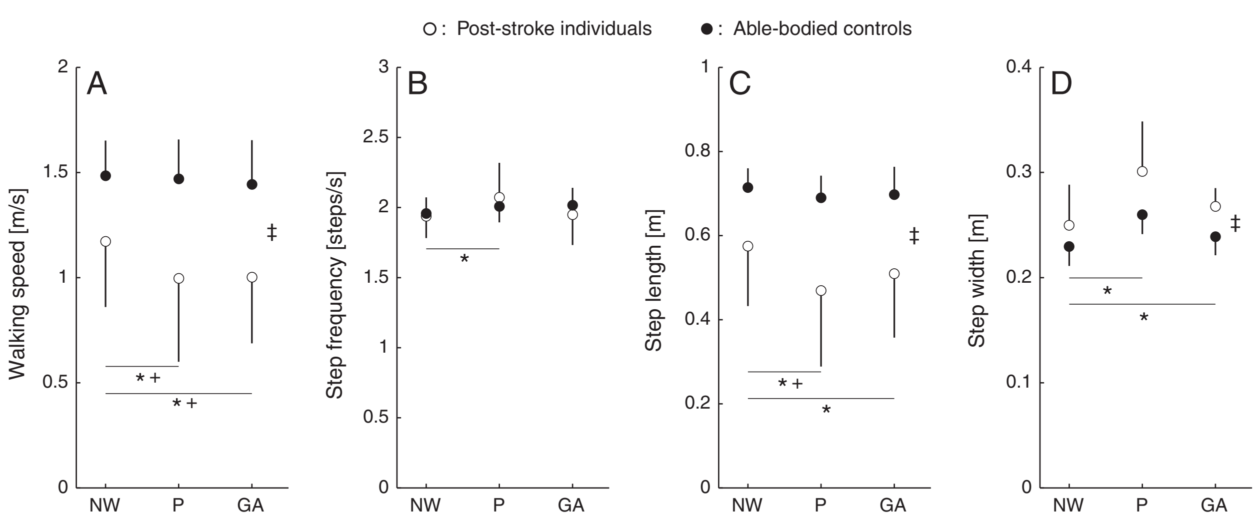

Fig. 3. Average and standard deviation of walking speed (A), step frequency (B), step length (C), and step width (D) for post-stroke individuals (n = 9) and healthy controls (n = 9). NW: normal walking; P: perturbation; GA: gait adaptability task. Significant group effects are indicated with ‡. Significant contrasts between normal walking and perturbed walking and/or between normal walking and walking with GA-task are indicated with * and significant group × manipulation interaction effects are indicated with +.

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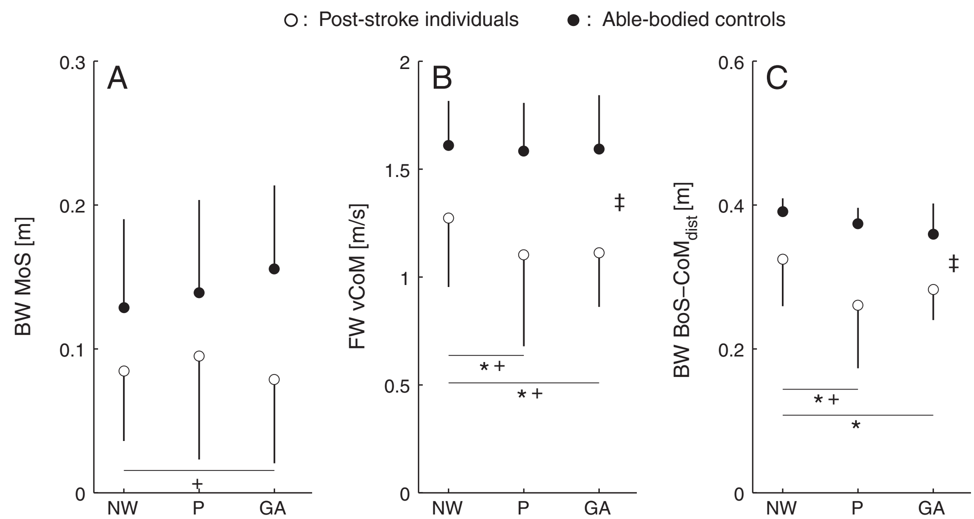

Fig. 5. Average and standard deviation of BW MoS (A) BW BoS-CoM$_{dist}$ (B) and FW vCoM (C) for post-stroke individuals (n = 9) and healthy controls (n = 9). NW: normal walking; P: perturbation; GA: gait adaptability task. Significant group effects are indicated with ‡. Significant contrasts between normal walking and perturbed walking and/or between normal walking and walking with GA-task are indicated with * and significant group × manipulation interaction effects are indicated with +.

perturbation we looked at the average adaptations of the gait pattern. For the trial with the GA-task we were interested in the anticipatory adaptations in the steady state gait pattern to facilitate the fast and accurate response necessary to hit the targets, and therefore we removed the strides in which the targets had to be hit. Post-stroke participants walked with comparable ML MoS compared to able-bodied participants, and were able to increase their ML MoS to the same degree as able-bodied participants in response to both manipulations of gait stability and gait adaptability, which was in contrast with our hypothesis. With respect to the BW MoS, these margins were smaller compared to the able-bodied group. This was especially the case for the condition with the GA-task, which appeared to be a challenging task for the post-stroke participants given both a smaller BW MoS and a reduced accuracy of the knee movements compared to the able-bodied controls, which was in line with our hypothesis. Although post-stroke participants did not actually fall during the experiment, both reduced backward margins of stability and reduced movement accuracy would bring post-stroke individuals at a higher risk of disrupting forward progression and possibly falling in situations during normal live walking in which fast and accurate adaptations of gait pattern are required.

In the current study, we found an overall larger excursion of the ML XCoM in the post-stroke group compared to the able-bodied controls, indicating larger ML body sway. In concert with that we found that post-stroke individuals walked with a larger step width and, considering their relatively low walking speed, a relatively high step frequency, compared to able-bodied participants. Both increases in

Table 2
Results for statistical analyses.

Parameter	Perturbation contrast	GA-task contrast	Main effect group	Group * perturbation contrast	Group * GA-task contrast
Step length	F = 50.675	F = 9.684	F = 11.300	F = 20.204	F = 3.520
	P < 0.01[a]	P < 0.01[a]	P < 0.01[a]	P < 0.01[a]	P = 0.079
	df = 1,16	df = 1,16	df = 1,16	df = 1,16	df = 1,16
Step width	F = 69.631	F = 6.898	F = 5.443	F = 1.450	F = 1.885
	P < 0.01[a]	P = 0.018[a]	P = 0.033[a]	P = 0.246	P = 0.189
	df = 1,16	df = 1,16	df = 1,16	df = 1,16	df = 1,16
Step frequency	F = 15.499	F = 4.287	F < 0.01	F = 3.497	F = 1.789
	P < 0.01[a]	P = 0.055	P = 0.930	P = 0.080	P = 0.200
	df = 1,16	df = 1,16	df = 1,16	df = 1,16	df = 1,16
Walking speed	F = 20.628	F = 12.657	F = 10.444	F = 14.407	F = 4.684
	P < 0.01[a]	P < 0.01[a]	P < 0.01[a]	P < 0.01[a]	P = 0.046[a]
	df = 1,16	df = 1,16	df = 1,16	df = 1,16	df = 1,16
ML MoS	F = 31.605	F = 1.348	F = 0.465	F = 0.154	F = 0.196
	P < 0.01[a]	P = 0.263	P = 0.505	P = 0.700	P = 0.664
	df = 1,16	df = 1,16	df = 1,16	df = 1,16	df = 1,16
BW MoS	F = 2.045	F = 4.337	F = 3.571	F < 0.01	F = 6.115
	P = 0.172	P = 0.054	P = 0.077	P = 0.989	P = 0.025[a]
	df = 1,16	df = 1,16	df = 1,16	df = 1,16	df = 1,16
XCoM$_{disp}$	F = 47.317	F = 6.823	F = 5.005	F = 0.237	F = 0.666
	P < 0.01[a]	P = 0.019[a]	P = 0.040[a]	P = 0.633	P = 0.426
	df = 1,16	df = 1,16	df = 1,16	df = 1,16	df = 1,16
BW BoS-CoM$_{dist}$	F = 70.345	F = 12.759	F = 10.082	F = 9.148	F = 4.551
	P < 0.01[a]	P < 0.01[a]	P < 0.01[a]	P < 0.01[a]	P = 0.049[a]
	df = 1,16	df = 1,16	df = 1,16	df = 1,16	df = 1,16
FW vCoM	F = 15.359	F = 9.883	F = 11.416	F = 25.749	F = 0.718
	P < 0.01[a]	P < 0.01[a]	P < 0.01[a]	P < 0.01[a]	P = 0.409
	df = 1,16	df = 1,16	df = 1,16	df = 1,16	df = 1,16

[a] Significant at the 0.05 level.

step width and frequency have been shown to enhance ML MoS (Hof, 2008; Hof et al., 2007). Therefore it seems valid to conclude that these gait adaptation allow post-stroke individuals to maintain a ML MoS of the same size, compared to able-bodied subjects, despite the larger excursion of the ML XCoM in this group (Hof, 2008; Hof et al., 2007). It could be argued that the larger step width might be the cause of the larger body sway instead of a compensatory strategy to deal with this larger sway. However, from the mechanical model introduced by Hof (Hof, 2008) it can be derived that an increase in step width will result in a net increase of the ML MoS despite a concomitant increase in body sway. Therefore increasing step width should likely be regarded as a strategy to deal with the consequences of increased ML sway and not only the cause of this increased sway.

For both groups ML excursions of the XCoM were larger during both the manipulations of gait stability and adaptability. In response to the platform perturbation post-stroke participants were able to withstand this manipulation, by increasing their ML MoS to the same degree as able-bodied participants, by further increasing step width and step frequency. Similarly, for the condition with the GA-task, the regulation of the ML MoS did not differ between both groups. Both groups were able to maintain their ML MoS, despite the disturbing effect of the manipulation, by increasing their step width. In agreement with earlier studies an increase in step frequency was absent in both groups (Hak et al., 2013a, 2013b), probably to prevent a decrease in available response time, which might affect the accuracy of the knee movement needed to hit the targets (Fitts, 1954). These findings of perturbed and unperturbed walking corroborate the notion that increasing step width and step frequency are functional gait adaptations that can be and are used by post-stroke individuals to preserve ML MoS.

In contrast with the regulation of the ML MoS, differences in the regulation of the BW MoS were observed between post-stroke participants and able-bodied participants. The BW MoS tended to be smaller for post-stroke participants compared to controls, especially for the trial in which the GA-task had to be performed. In response to the GA-task, able-bodied participants increased their BW MoS by reducing step length without a concomitant reduction in walking speed. Post-stroke participants also did reduce step length in response to the GA-task, but this was accompanied by a reduction of walking speed. BW MoS can be increased by either decreasing step length (reducing the distance between CoM and BoS at initial contact) or by increasing forward velocity at initial contact (Espy et al., 2010a; Pai & Patton, 1997). The results shown in Fig. 5 demonstrate that the distance between the CoM and the leading foot at initial contact was indeed smaller for post-stroke participants as a result of the smaller step length. However, forward velocity of the CoM at initial contact was considerably lower compared to able-bodied controls, especially during the trial with the GA-task. In contrast to the general notion that a lower walking speed might be a strategy to increase gait stability (Dingwell & Marin, 2006; England & Granata, 2007; Kang & Dingwell, 2008; Krasovsky et al., 2012), these results demonstrate that the lower walking speed in post-stroke individuals seems to cause a decrease in the BW MoS, and might therefore increase the risk of a disruption of forward progression and possibly a backward fall. It remains speculative why post-stroke individuals have problems with the regulation of the BW MoS. Possibly, post-stroke individuals experience problems with selecting an appropriate combination of step length and frequency, at a given walking speed, as evidenced by the seemingly inappropriate reduction in step length and speed in response to the GA-task. This lack of adaptation might have a physical cause (i.e. impaired movement selectivity or a reduced push-off capacity of the hemiparetic leg (Balasubramanian et al., 2007; Neptune & McGowan, 2011; Roerdink & Beek, 2011; Turns et al., 2007)) or a mental cause (a higher fear of falling (Maki, 1997) or a conflict between the cognitive demands of the GA-task and walking ability (Hyndman et al., 2006; Plummer-D'Amato et al., 2008; Smulders et al., 2012)). From the data in the current study this cannot be resolved.

A limitation of this study is the estimation of the CoM as the average of the markers attached to the LASI, RASI, LPSI, and RPSI of the pelvis to calculate the XCoM. It is conceivable that the post-stroke group has a larger sway of the upper body, in which case this method could cause an underestimation of the displacement of the XCoM, and therefore an overestimation of the size of the MoS. However, from a study of De Bujunda et al. (De Bujunda et al., 2004) it appeared that, at least in the frontal plane, displacements and accelerations of the shoulders and pelvis during walking are very similar, and this was the case for both post-stroke individuals and able-bodied controls. Therefore, we expect that the potential error made in the estimation of the CoM position in the current study will be small. Besides, this error likely affects only the overall group difference in the MoS, and not the adaptation of these measures in response to the manipulations.

When interpreting the results of the present study, it has to be taken into account that the post-stroke individuals that participated in this study were all relatively good walkers. This may explain why the post-stroke subjects were able to walk with comparable ML MoS as the able-bodied controls and why all subjects, except one, were able to complete the protocol without actually falling. Therefore generalization of the results to subjects with a more severe hemiparesis needs to be done with caution.

Secondly, it is of importance to mention that the calculation of the MoS as a measure of stability is based on an inverted pendulum model, which is a strong simplification of human walking. This model is designed to quantify the contribution of a change in foot placement, like an increase in step width, in the regulation of dynamic balance (Hof, 2008). However, wider steps will also result in larger angular and linear momenta of the pendulum at foot contact. This is trivial for an inverted pendulum, because its legs are rigid, but in humans larger linear and angular momenta at foot placement require larger joint moments in the 'new' stance leg to reduce and reverse these momenta. This aspect of balance control is ignored in the model. The results of the current study have shown that the relatively large step width in post-stroke individuals results in a proper foot placement with respect to the XCoM. However, generating sufficient joint moments to compensate for the increased momenta might be a problem in post-stroke individuals, especially in view of associated muscle weakness. Controlling the size of the MoS is just one prerequisite for preventing falls during walking, and consequently the translation of the results for the MoS to the more general concept of fall risk should be done with care.

In conclusion, despite the larger ML body sway, post-stroke participants were able to regulate their ML MoS, during all experimental conditions, to the same degree as the able-bodied subjects. However this required a larger step width and a relatively high step frequency compared to able-bodied participants. BW MoS tended to be smaller for post-stroke participants, compared to able-bodied participants, especially for the condition in which the GA-task had to be performed. In response to the GA-task, BW MoS decreased for the post-stroke group, while able-bodied participants were able to maintain their BW MoS. An explanation for these smaller BW MoS seems to be that post-stroke participants significantly decreased their walking speed, while able-bodied participants maintained their walking speed in response to both manipulations through an effective adaptation of both step length and step frequency. Future studies should aim on identifying whether post-stroke individuals are really limited in selecting different step length and frequency combinations at a constant walking speed and which impairments cause these limitations.

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